Estimation of patient-specific mechanical parameters in pulmonary diseases

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1. Introduction

Pulmonary diseases are among the main causes of death in the world, thus representing an important global health burden. Among them, interstitial pulmonary diseases affect pulmonary tissue and its alveolar structure, thus highly impacting the pulmonary function. Idiopathic Pulmonary Fibrosis (IPF), for instance, is a chronic disease, in which collagen fibers accumulate into interstitial tissue, leading to thickening, stiffening and damage of alveolar walls. This disease remains poorly understood, poorly diagnosed and poorly treated and represents a real clinical challenge [5]. Mechanical modeling-based tools, in interaction with data such as medical imaging, could help clinicians in classifying patients and thus deciding on the treatment options.

2. Method

2.1 Lung poromechanical model

The lungs are porous organs, composed of tissue, air and blood. We recently developed a model of the lungs at the breathing time scale and the organ space scale [6], based on a general poromechanical formulation compatible with large strains and thermodynamics [1], where the "solid" phase is composed of both tissue and blood while the fluid phase is the air. Several pulmonary-specific hypotheses have been formulated, assuming that the transformation is quasi-static and that the fluid pressure inside the alveoli is homogeneous and given. The problem is then simplified and becomes a coupled problem between the two unknowns, the displacement $\underline{U}$ (or equivalently the deformation mapping $\underline{\Phi}$) and the porosity ϕ.

In our model, specific boundary conditions are used to model loadings applied to the lungs: a pressure on lung surface representing pleural pressure, and a frictionless contact with the moving thorax [6]. Moreover, the constitutive behavior of our model allows to reproduce the volumetric response of lungs to a change of pressure as observed in experimental data [4], as well as the quasi-incompressibility of the solid phase [6].

2.2 Model personalization

The proposed model can be personalized using clinical data. Two 3D CT-scans I_0 and I_1, acquired at end-exhalation and at end-inhalation respectively, are used (1) to generate (after segmentation [2]) a patient-specific geometry of the lungs (volume mesh) and thorax (surface mesh), (2) to get a personalized porosity field, and (3) to compute lung and thorax displacement field by DIC (Digital Image Correlation) [3]. In principle, pleural pressure could be measured in patients and used in the model. However, in this work, normal values were used since the data was not available. As a consequence, the results obtained are relative to these values.

Regional mechanical parameters θ are estimated by minimizing a cost function f describing the error between data and model. For a given set of parameters, the unloaded configuration associated with the measured end-exhalation configuration is first computed through the resolution of an inverse elastostatic problem. Then, the end-inhalation configuration is computed, and compared to the measured one.

We investigated two types of parametrization of the problem, where the estimated material parameters can be the absolute (*i.e.*, independent from porosity, in which case the effective parameters depend linearly on the porosity) or the effective (*i.e.*, those characterizing the mixture) parameters.

We also investigated two different cost functions, differing by the nature of the data considered. The first one is based on the lung displacement field computed by image registration:

$$f_{\text{DIC}}(\theta) = \frac{\|\underline{U}_{\text{model}}(\theta) - \underline{U}_{\text{DIC}}\|_{L^2}}{\|\underline{U}_{\text{DIC}}\|_{L^2}}.$$

Note that special care must be taken for pulmonary image registration because of the large discontinuity in

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the displacement field caused by the lung-thorax sliding. The second cost function, which corresponds to integrated image correlation approaches, is based on the 3DCT images directly and contains both a shape term and a configuration term:

$$f_{\text{image}}(\theta) = \frac{\| I_1 \circ \underline{\Phi}_{\text{model}}(\theta) - I_0 \|_{L^2}}{\| I_0 \|_{L^2}} + k\, DICE(\theta),$$

where *DICE* is the Dice score of the computed *vs.* measured end-inhalation configurations and k weighs the shape and configuration terms.

3. Results and discussion

The estimation pipeline was run on healthy controls as well as IPF patients.

To prove the classification potential of the estimation, mechanical parameters are estimated in three different cases: considering the lung as one homogeneous zone (case *1z*); as two homogeneous zones defined arbitrarily (case *2zarb*); as two homogeneous zones defined according to the segmentation of the disease by a radiologist (case *2zseg*). For both cost functions, we obtained: $f_{1z} > f_{2zarb} > f_{2zseg}$. Notably, the cost function f_{DIC} improves from 23.5 % and 20.6 % to 18.3 % (which correspond to 6.3 mm, 5.6 mm and 4.9 mm in terms of displacement error), for the cases *1z*, *2zarb*, *2zseg* respectively. These results show that the model including physiological segmentation of the disease better represents the data. Statistical significance will now be evaluated.

We also studied the impact of estimating the effective *vs.* absolute material parameters. In all cases, the model with absolute parameters performs better than with effective parameters. The displacement error decreases from 5.1 mm with effective parameters to 4.9 mm with absolute parameters in the case *2zseg* using f_{DIC}. Moreover, for the case *2zseg*, both cost functions result in a fibrosis zone stiffer than the healthy zone, which is consistent with the fact that fibrosis leads to tissue stiffening. This indicates that using a poromechanical model, thus taking into account porosity, should improve classification potential.

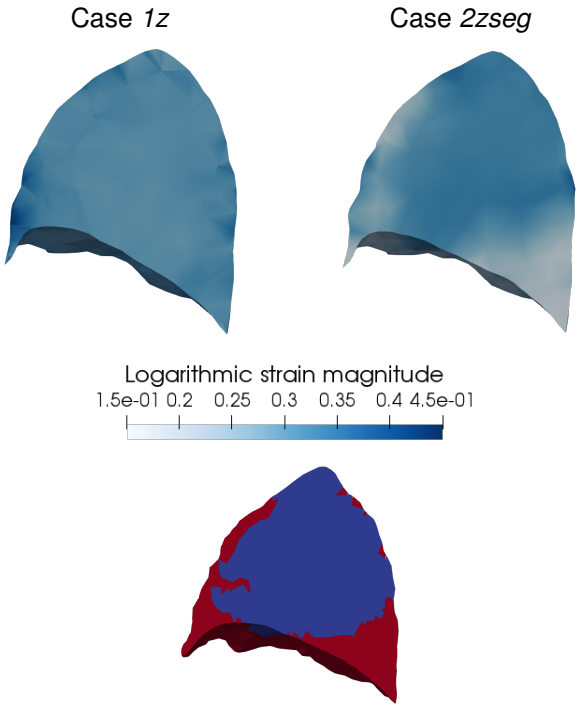

Figure 1: (Top) Logarithmic strain magnitude in a sagittal cross-section at end-inhalation for the case *1z* and the case *2zseg*. (Bottom) Visualization of the fibrosis segmentation in the same sagittal cross-section: healthy part in blue, fibrotic part in red.

4. Conclusion

We developed a pulmonary poromechanical model, which takes into account lung porosity and which can be personalized using clinical data acquired in routine. The model can be applied on complex pathological cases involving diseases with an impact on lung mechanical behavior, like fibrosis or emphysema. It could then be used as an augmented diagnosis tool to quantify the mechanical changes induced by the disease, thus providing objective and quantitative classification tools to clinicians.

5. References

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